

IMAGE CROPPING FOR ASYMMETRICAL IMAGING

by

**Vincent S. Polkus
Thomas M. Leeds
Mohamed Ali Hamadeh
Ping Xue**

EXPRESS MAIL MAILING LABEL	
NUMBER:	EL 652 334 796 US
DATE OF DEPOSIT:	January 5, 2001
<i>Pursuant to 37 C.F.R. § 1.10, I hereby certify that I am personally depositing this paper or fee with the U.S. Postal Service, "Express Mail Post Office to Addressee" service on the date indicated above in a sealed envelope (a) having the above-numbered Express Mail label and sufficient postage affixed, and (b) addressed to the Assistant Commissioner for Patents, Washington, D.C. 20231.</i>	
01/05/01 <i>Lynda Howell</i> Date	<i>Lynda Howell</i> Lynda Howell

IMAGE CROPPING FOR ASYMMETRICAL IMAGING

FIELD OF THE INVENTION

The present invention relates generally to digital imaging systems, such as
5 digital x-ray imaging systems. More particularly, the invention relates to a technique for cropping or refining image data for more efficient processing when an imaging system is set to create an image off center from a radiation source location.

10

BACKGROUND OF THE INVENTION

15

20

Digital x-ray imaging systems are becoming increasingly widespread for producing digital data which can be reconstructed into useful radiographic images. In current digital x-ray imaging systems, radiation from a source is directed toward a subject, typically a patient in a medical diagnostic application. A portion of the radiation passes through the patient and impacts a detector. The surface of the detector converts the radiation to light photons which are sensed. The detector is divided into a matrix of discrete picture elements or pixels, and encodes output signals based upon the quantity or intensity of the radiation impacting each pixel region. Because the radiation intensity is altered as the radiation passes through the patient, the images reconstructed based upon the output signals provide a projection of the patient's tissues similar to those available through conventional photographic film techniques.

25

30

Digital x-ray imaging systems are particularly useful due to their ability to collect digital data which can be reconstructed into the images required by radiologists and diagnosing physicians, and stored digitally or archived until needed. In conventional film-based radiography techniques, actual films were prepared, exposed, developed and stored for use by the radiologist. While the films provide an excellent diagnostic tool, particularly due to their ability to capture significant anatomical detail, they are inherently difficult to transmit between locations, such as from an imaging facility or department to various

physician locations. The digital data produced by direct digital x-ray systems, on the other hand, can be processed and enhanced, stored, transmitted via networks, and used to reconstruct images which can be displayed on monitors and other soft copy displays at any desired location. Similar advantages are offered by digitizing systems which convert conventional radiographic images from film to digital data.

In certain type of imaging systems, such as digital x-ray systems, the radiation source may be positioned at various locations along an imaging area, with the detector typically being positioned at a corresponding location. For example, the source and detector may be moved along a longitudinal centerline of a patient support and, in certain systems, in a direction transverse to the centerline. Such positioning is useful for imaging specific anatomies or limbs, while exposing a patient to a minimal level of radiation.

In digital imaging systems, the computational load imposed on the image data processing circuitry is related to the amount of information collected. For larger or higher resolution images, or images employing a greater dynamic range for each pixel, significant quantities of data may be collected and processed to obtain the final data set used to reconstruct the image. Where smaller areas are imaged, such as specific anatomies in x-ray systems, electronic cropping may be used to reduce the total amount of data collected or processed. In general, such cropping entails selectively sampling or processing data from those pixels corresponding to the desired image area, the remaining pixels being considered to contain little or no useful information.

For asymmetrical imaging (i.e. where the source is angularly positioned with respect to a projection line through the source and orthogonal to the detector and/or where the image center is not coincident with the detector center), no effective automated digital cropping technique has been developed. Consequently, in such cases, an operator or clinician may be required manually to view the image and crop the data after processing.

5

There is a need, therefore, for an improved image data cropping technique which will allow for automated digital cropping of image data in asymmetrical imaging situations. There is a particular need for an approach which permits the quantity of data sampled or processed to be reduced and which reduces the need for clinicians to manually view and crop resulting images.

SUMMARY OF THE INVENTION

10

The present invention provides a technique for cropping image data designed to respond to these needs. The technique may be used in newly-installed systems, but may be easily retrofitted into existing systems to enhance performance and reduce computational loads. While the technique may be used in a wide range of system types, it is particularly well-suited to digital x-ray systems which permit asymmetrical imaging by translating and/or tilting (rotating) a radiation source assembly, or by swiveling the associated beam collimation device with respect to an imaging plane.

15

The technique allows for determining the projection area of an imaging beam, such as an x-ray beam, on the basis of system geometry. In particular, the technique may take into account various spatial and angular positions and rotational positions of both a radiation source and a collimator, which, in combination, may result in various forms and orientations of an imaging projection on the image plane. In a full implementation, angular translation and rotation can be implemented for both the source and the collimator in three independent directions in space. More limited implementations may be envisioned, in which angular position and rotation are more limited, such as angular position of a source with respect to a centerline of the system only. The technique also allows for determining whether the projection area falls within a detector framework or boundaries. Thus, in addition to limiting the amount of data sampled, and improving the computational efficiency of the system, the technique may be used to avoid unnecessary exposure to radiation, or to limit radiation exposure to a useful area of a detector.

20

25

30

The algorithm also includes the condition where the x-ray beam is not angulated but rather offset with respect to a condition where the beam and detector centers are coincident. This specific case is possible when the x-ray field size is less than that of the detector and the central ray is orthogonal to the detector plane. In
5 this situation, the operator is able to displace the source assembly with respect to the detector center and still produce an acceptable diagnostic image that fully captures the anatomical features of the region of interest. (This is possible by comparing the predicted coordinates of the x-ray beam impingement points/vertices with the coordinates of the detector edge and cropping the image if those coordinates are
10 contained in the space occupied by the image detector. The algorithm also accommodates beam angulation in the generalized formulation. When there is no beam angulation, diagonal terms of the transformation matrices become 1 (cosine (0)).
15

BRIEF DESCRIPTION OF THE DRAWINGS

Fig. 1 is a diagrammatical overview of a digital x-ray imaging system in which the present technique is incorporated;

Fig. 2 is a diagrammatical representation of certain of the functional circuitry for producing image data in a detector of the system of Fig. 1 to produce
20 image data for reconstruction;

Fig. 3 is a partial sectional view illustrating an exemplary detector structure for producing the image data;

Fig. 4 is a diagrammatical perspective view of an x-ray imaging system of the type shown in Fig. 1, and illustrating ranges of movement of a source and
25 detector along an imaging area;

Fig. 5 is a plan view of image areas of the type obtainable through proper adjustment of the arrangement of Fig. 4;

Fig. 6 is a diagrammatical illustration of a coordinate system for movement of a radiation source in the arrangement of Fig. 4;

30 Fig. 7 is a diagrammatical illustration of a coordinate system for movement of a collimator in the arrangement of Fig. 4;

Fig. 8 is an elevational view of an x-ray beam from a source directed to a detector in an exemplary asymmetrical imaging application;

Fig. 9 is a plan view of the same beam; and

Fig. 10 is a flow chart illustrating exemplary logic in performing digital cropping of an image taken in an asymmetrical imaging application.

5

DETAILED DESCRIPTION OF THE INVENTION

Fig. 1 illustrates diagrammatically an imaging system 10 for acquiring and processing discrete pixel image data. In the illustrated embodiment, system 10 is a digital x-ray system designed both to acquire original image data, and to process the image data for display in accordance with the present technique. In the embodiment illustrated in Fig. 1, imaging system 10 includes a source of x-ray radiation 12 positioned adjacent to a collimator 14. Collimator 14 permits a stream of radiation 16 to pass into a region in which a subject, such as a human patient 18 is positioned. 10 A portion of the radiation 20 passes through or around the subject and impacts a digital x-ray detector, represented generally at reference numeral 22. As described more fully below, detector 22 converts the x-ray photons received on its surface to lower energy photons, and subsequently to electric signals which are acquired and processed to reconstruct an image of the features within the subject.

15

Source 12 is controlled by a power supply/control circuit 24 which furnishes both power and control signals for examination sequences. Moreover, detector 22 is coupled to a detector controller 26 which commands acquisition of the signals generated in the detector. Detector controller 26 may also execute various signal processing and filtration functions, such as for initial adjustment of dynamic ranges, interleaving of digital image data, and so forth. Both power supply/control circuit 24 and detector controller 26 are responsive to signals from a system controller 28. 20 In general, system controller 28 commands operation of the imaging system to execute examination protocols and to process acquired image data. In the present context, system controller 28 also includes signal processing circuitry, typically based upon a general purpose or application-specific digital computer, associated

25

30

memory circuitry for storing programs and routines executed by the computer, as well as configuration parameters and image data, interface circuits, and so forth.

In the embodiment illustrated in Fig. 1, system controller 28 is linked to at least one output device, such as a display or printer as indicated at reference numeral 30. The output device may include standard or special purpose computer monitors and associated processing circuitry. One or more operator workstations 32 may be further linked in the system for outputting system parameters, requesting examinations, viewing images, and so forth. In general, displays, printers, workstations, and similar devices supplied within the system may be local to the data acquisition components, or may be remote from these components, such as elsewhere within an institution or hospital, or in an entirely different location, linked to the image acquisition system via one or more configurable networks, such as the Internet, virtual private networks, and so forth.

Fig. 2 is a diagrammatical representation of functional components of digital detector 22. Fig. 2 also represents an imaging detector controller or IDC 34 which will typically be configured within detector controller 26. IDC 34 includes a CPU or digital signal processor, as well as memory circuits for commanding acquisition of sensed signals from the detector. IDC 34 is coupled via two-way fiberoptic conductors to detector control circuitry 36 within detector 22. IDC 34 thereby exchanges command signals for image data within the detector during operation.

Detector control circuitry 36 receives DC power from a power source, represented generally at reference numeral 38. Detector control circuitry 36 is configured to originate timing and control commands for row and column drivers used to transmit signals during data acquisition phases of operation of the system. Circuitry 36 therefore transmits power and control signals to reference/regulator circuitry 40, and receives digital image pixel data from circuitry 40.

In a presently preferred embodiment illustrated, detector 22 consists of a scintillator that converts x-ray photons received on the detector surface during

examinations to lower energy (light) photons. An array of photodetectors then converts the light photons to electrical signals which are representative of the number of photons or the intensity of radiation impacting individual pixel regions of the detector surface. Readout electronics convert the resulting analog signals to digital values that can be processed, stored, and displayed, such as in a display 30 or a workstation 32 following reconstruction of the image. In a present form, the array of photodetectors is formed on a single base of amorphous silicon. The array elements are organized in rows and columns, with each element consisting of a photodiode and a thin film transistor. The cathode of each diode is connected to the source of the transistor, and the anodes of all diodes are connected to a negative bias voltage. The gates of the transistors in each row are connected together and the row electrodes are connected to the scanning electronics. The drains of the transistors in a column are connected together and an electrode of each column is connected to readout electronics.

In the particular embodiment illustrated in Fig. 2, by way of example, a row bus 42 includes a plurality of conductors for enabling readout from various columns of the detector, as well as for disabling rows and applying a charge compensation voltage to selected rows, where desired. A column bus 44 includes additional conductors for commanding readout from the columns while the rows are sequentially enabled. Row bus 42 is coupled to a series of row drivers 46, each of which commands enabling of a series of rows in the detector. Similarly, readout electronics 48 are coupled to column bus 44 for commanding readout of all columns of the detector.

In the illustrated embodiment, row drivers 46 and readout electronics 48 are coupled to a detector panel 50 which may be subdivided into a plurality of sections 52. Each section 52 is coupled to one of the row drivers 46, and includes a number of rows. Similarly, each column driver 48 is coupled to a series of columns. The photodiode and thin film transistor arrangement mentioned above thereby define a series of pixels or discrete picture elements 54 which are arranged in rows 56 and

columns 58. The rows and columns define an image matrix 60, having a height 62 and a width 64.

As also illustrated in Fig. 2, each pixel 54 is generally defined at a row and column crossing, at which a column electrode 68 crosses a row electrode 70. As mentioned above, a thin film transistor 72 is provided at each crossing location for each pixel, as is a photodiode 74. As each row is enabled by row drivers 46, signals from each photodiode may be accessed via readout electronics 48, and converted to digital signals for subsequent processing and image reconstruction.

10

Fig. 3 generally represents an exemplary physical arrangement of the components illustrated diagrammatically in Fig. 2. As shown in Fig. 3, the detector may include a glass substrate 76 on which the components described below are disposed. Column electrodes 68 and row electrodes 70 are provided on the substrate, and an amorphous silicon flat panel array 78 is defined, including the thin film transistors and photodiodes described above. A scintillator 80 is provided over the amorphous silicon array for receiving radiation during examination sequences as described above. Contact fingers 82 are formed for communicating signals to and from the column and row electrodes, and contact leads 84 are provided for communicating the signals between the contact fingers and external circuitry.

15

20

The foregoing exemplary imaging system may permit various types of alternative positioning of the x-ray source, the beam collimator, and the detector to allow for considerable latitude in selecting a segment of a subject which will be imaged. Fig. 4 illustrates ranges of movement of a source and a detector along an imaging area. As shown in Fig. 4, the source 12 projects a beam 16 of radiation through collimator 14 toward the detector 22. In the system arrangement of Fig. 4, designated generally by the reference numeral 86, an effective point source 88 of radiation is contained within the source 12 and is projected through an aperture 90 in the collimator. In the illustrated example, aperture 90 is square or rectangular, lending to the beam a similar shape, as defined by straight lines joining points or vertices 92, 94, 96 and 98. As will be appreciated by those skilled in the art, in

25

30

practice, the beam may radiate well beyond the bounds of aperture 90, with the aperture serving to direct a portion of the beam to the detector within the desired imaging area.

5 An imaging or impingement plane 100 generally corresponds to a surface on which or adjacent to which a patient is positioned in a medical diagnostic imaging application. In other contexts, the plane may serve as support or background for other types of objects. An impingement or imaging area 102 is defined where the beam 16, shaped by the collimator aperture 90, impinges the plane 100. To permit
10 selection of the region along plane 100 to be imaged, the source 12 may be moved as indicated by reference numeral 104, along a range of movement 106. In the illustration of Fig. 4, the range 106 is generally along or parallel to a longitudinal centerline of the plane 100. Similar movement 108 is available for the detector 22, along a range of movement 110. In general, collimator 14 is displaced with the
15 radiation source, the two forming a radiation source assembly, although rotational and angular positioning of the collimator may be provided independent of the rotational or angular position of the source and described below.

Exemplary projections of beam 16 in plane 100 of the arrangement of Fig. 4
20 are illustrated in Fig. 5. As shown in Fig. 5, limits 112 and 114 on the movement of the detector 22 will typically impose corresponding limits on the range of imaging areas available. However, where the source and/or collimator may be moved along a further range and angularly positioned with respect to a centerline 116 of the system, various projection areas may be obtained. For example, asymmetrical
25 imaging applications may permit projections 118 and 120 to be formed off of centerline 116. While in the simplified illustration of Fig. 5, these areas have generally rectangular shapes, in practice, their shapes may be manipulated by appropriate angular positioning and rotation of the source and/or collimator as described more fully below.

30

As will be appreciated by those skilled in the art, projections off of centerline 116 may be extremely useful in reducing radiation exposure, while allowing for

imaging of selected anatomies or features of a patient or subject. Where a detector 22 provides an area substantially larger than the image area, however, sampling of the entire detector, and processing of image data from the entire detector matrix can result in substantial computational loads. To provide greater computational efficiency, therefore, the present technique provides for selective processing or "digital cropping" of the image data in such applications.

In the analysis of the image for determining a projection area for cropping, reference may be made to coordinate systems for angular positioning and rotation of both the radiation source and the collimator. Figs. 6 and 7 illustrate various degrees of freedom of motion of a source and a collimator, respectively, in a system of the type illustrated in Fig. 4. In particular, Fig. 6 represents a coordinate system 122 in which the radiation source may be translated angularly, and rotated. As noted above, the beam from the point source 88 is formed or shaped by the opening in the collimator, such as corresponding to vertices 92, 94, 96 and 98 and the edges extending therebetween. Where the system permits, however, the source may be positioned with respect to mutually orthogonal coordinate axes, such as an X1 axis 124, an X2 axis 126, and an X3 axis 128. In addition to angular positioning within this coordinate system 122, the source may be provided with freedom of rotational movement about the axis or any combination of the axes. Resulting projection lines extending through the corners of the collimator aperture are thereby defined as indicated by lines 130, 132, 134 and 136 in Fig. 6. Moreover, a centerline passing through the midpoint of the collimator is defined as indicated at reference numeral 138. Again, it should be noted that the lines illustrated in Fig. 6 are simply projections of the beam limits following collimation, and do not necessarily correspond to the shape of the beam prior to impingement with the collimator.

In certain systems and where desired, similar freedom of movement may be provided for the collimator to allow further shaping of the image area. As illustrated in Fig. 7, with respect to the point source 88, the collimator coordinate system 140 may provide for translation along and rotation about additional mutual orthogonal axes, including a Y1 axis 142, a Y2 axis 144 and a Y3 axis 146. Again, the lines of

10
20
30
40
50
60
70
80
90
100
110
120
130
140
150
160
170
180
190
200
210
220
230
240
250
260
270
280
290
300
310
320
330
340
350
360
370
380
390
400
410
420
430
440
450
460
470
480
490
500
510
520
530
540
550
560
570
580
590
600
610
620
630
640
650
660
670
680
690
700
710
720
730
740
750
760
770
780
790
800
810
820
830
840
850
860
870
880
890
900
910
920
930
940
950
960
970
980
990
1000
1010
1020
1030
1040
1050
1060
1070
1080
1090
1100
1110
1120
1130
1140
1150
1160
1170
1180
1190
1200
1210
1220
1230
1240
1250
1260
1270
1280
1290
1300
1310
1320
1330
1340
1350
1360
1370
1380
1390
1400
1410
1420
1430
1440
1450
1460
1470
1480
1490
1500
1510
1520
1530
1540
1550
1560
1570
1580
1590
1600
1610
1620
1630
1640
1650
1660
1670
1680
1690
1700
1710
1720
1730
1740
1750
1760
1770
1780
1790
1800
1810
1820
1830
1840
1850
1860
1870
1880
1890
1900
1910
1920
1930
1940
1950
1960
1970
1980
1990
2000
2010
2020
2030
2040
2050
2060
2070
2080
2090
2100
2110
2120
2130
2140
2150
2160
2170
2180
2190
2200
2210
2220
2230
2240
2250
2260
2270
2280
2290
2200
2210
2220
2230
2240
2250
2260
2270
2280
2290
2300
2310
2320
2330
2340
2350
2360
2370
2380
2390
2400
2410
2420
2430
2440
2450
2460
2470
2480
2490
2500
2510
2520
2530
2540
2550
2560
2570
2580
2590
2600
2610
2620
2630
2640
2650
2660
2670
2680
2690
2700
2710
2720
2730
2740
2750
2760
2770
2780
2790
2800
2810
2820
2830
2840
2850
2860
2870
2880
2890
2900
2910
2920
2930
2940
2950
2960
2970
2980
2990
3000
3010
3020
3030
3040
3050
3060
3070
3080
3090
3100
3110
3120
3130
3140
3150
3160
3170
3180
3190
3200
3210
3220
3230
3240
3250
3260
3270
3280
3290
3300
3310
3320
3330
3340
3350
3360
3370
3380
3390
3400
3410
3420
3430
3440
3450
3460
3470
3480
3490
3500
3510
3520
3530
3540
3550
3560
3570
3580
3590
3600
3610
3620
3630
3640
3650
3660
3670
3680
3690
3700
3710
3720
3730
3740
3750
3760
3770
3780
3790
3800
3810
3820
3830
3840
3850
3860
3870
3880
3890
3900
3910
3920
3930
3940
3950
3960
3970
3980
3990
4000
4010
4020
4030
4040
4050
4060
4070
4080
4090
4000
4010
4020
4030
4040
4050
4060
4070
4080
4090
4100
4110
4120
4130
4140
4150
4160
4170
4180
4190
4200
4210
4220
4230
4240
4250
4260
4270
4280
4290
4300
4310
4320
4330
4340
4350
4360
4370
4380
4390
4400
4410
4420
4430
4440
4450
4460
4470
4480
4490
4500
4510
4520
4530
4540
4550
4560
4570
4580
4590
4600
4610
4620
4630
4640
4650
4660
4670
4680
4690
4700
4710
4720
4730
4740
4750
4760
4770
4780
4790
4800
4810
4820
4830
4840
4850
4860
4870
4880
4890
4900
4910
4920
4930
4940
4950
4960
4970
4980
4990
5000
5010
5020
5030
5040
5050
5060
5070
5080
5090
5000
5010
5020
5030
5040
5050
5060
5070
5080
5090
5100
5110
5120
5130
5140
5150
5160
5170
5180
5190
5200
5210
5220
5230
5240
5250
5260
5270
5280
5290
5300
5310
5320
5330
5340
5350
5360
5370
5380
5390
5400
5410
5420
5430
5440
5450
5460
5470
5480
5490
5500
5510
5520
5530
5540
5550
5560
5570
5580
5590
5600
5610
5620
5630
5640
5650
5660
5670
5680
5690
5700
5710
5720
5730
5740
5750
5760
5770
5780
5790
5800
5810
5820
5830
5840
5850
5860
5870
5880
5890
5900
5910
5920
5930
5940
5950
5960
5970
5980
5990
6000
6010
6020
6030
6040
6050
6060
6070
6080
6090
6000
6010
6020
6030
6040
6050
6060
6070
6080
6090
6100
6110
6120
6130
6140
6150
6160
6170
6180
6190
6200
6210
6220
6230
6240
6250
6260
6270
6280
6290
6300
6310
6320
6330
6340
6350
6360
6370
6380
6390
6400
6410
6420
6430
6440
6450
6460
6470
6480
6490
6500
6510
6520
6530
6540
6550
6560
6570
6580
6590
6600
6610
6620
6630
6640
6650
6660
6670
6680
6690
6700
6710
6720
6730
6740
6750
6760
6770
6780
6790
6800
6810
6820
6830
6840
6850
6860
6870
6880
6890
6900
6910
6920
6930
6940
6950
6960
6970
6980
6990
7000
7010
7020
7030
7040
7050
7060
7070
7080
7090
7000
7010
7020
7030
7040
7050
7060
7070
7080
7090
7100
7110
7120
7130
7140
7150
7160
7170
7180
7190
7200
7210
7220
7230
7240
7250
7260
7270
7280
7290
7300
7310
7320
7330
7340
7350
7360
7370
7380
7390
7400
7410
7420
7430
7440
7450
7460
7470
7480
7490
7500
7510
7520
7530
7540
7550
7560
7570
7580
7590
7600
7610
7620
7630
7640
7650
7660
7670
7680
7690
7700
7710
7720
7730
7740
7750
7760
7770
7780
7790
7800
7810
7820
7830
7840
7850
7860
7870
7880
7890
7900
7910
7920
7930
7940
7950
7960
7970
7980
7990
8000
8010
8020
8030
8040
8050
8060
8070
8080
8090
8000
8010
8020
8030
8040
8050
8060
8070
8080
8090
8100
8110
8120
8130
8140
8150
8160
8170
8180
8190
8200
8210
8220
8230
8240
8250
8260
8270
8280
8290
8300
8310
8320
8330
8340
8350
8360
8370
8380
8390
8400
8410
8420
8430
8440
8450
8460
8470
8480
8490
8500
8510
8520
8530
8540
8550
8560
8570
8580
8590
8600
8610
8620
8630
8640
8650
8660
8670
8680
8690
8700
8710
8720
8730
8740
8750
8760
8770
8780
8790
8800
8810
8820
8830
8840
8850
8860
8870
8880
8890
8900
8910
8920
8930
8940
8950
8960
8970
8980
8990
9000
9010
9020
9030
9040
9050
9060
9070
9080
9090
9100
9110
9120
9130
9140
9150
9160
9170
9180
9190
9200
9210
9220
9230
9240
9250
9260
9270
9280
9290
9300
9310
9320
9330
9340
9350
9360
9370
9380
9390
9400
9410
9420
9430
9440
9450
9460
9470
9480
9490
9500
9510
9520
9530
9540
9550
9560
9570
9580
9590
9600
9610
9620
9630
9640
9650
9660
9670
9680
9690
9700
9710
9720
9730
9740
9750
9760
9770
9780
9790
9800
9810
9820
9830
9840
9850
9860
9870
9880
9890
9900
9910
9920
9930
9940
9950
9960
9970
9980
9990
10000
10010
10020
10030
10040
10050
10060
10070
10080
10090
10000
10010
10020
10030
10040
10050
10060
10070
10080
10090
10100
10110
10120
10130
10140
10150
10160
10170
10180
10190
10200
10210
10220
10230
10240
10250
10260
10270
10280
10290
10300
10310
10320
10330
10340
10350
10360
10370
10380
10390
10400
10410
10420
10430
10440
10450
10460
10470
10480
10490
10500
10510
10520
10530
10540
10550
10560
10570
10580
10590
10600
10610
10620
10630
10640
10650
10660
10670
10680
10690
10700
10710
10720
10730
10740
10750
10760
10770
10780
10790
10800
10810
10820
10830
10840
10850
10860
10870
10880
10890
10900
10910
10920
10930
10940
10950
10960
10970
10980
10990
11000
11010
11020
11030
11040
11050
11060
11070
11080
11090
11100
11110
11120
11130
11140
11150
11160
11170
11180
11190
11200
11210
11220
11230
11240
11250
11260
11270
11280
11290
11300
11310
11320
11330
11340
11350
11360
11370
11380
11390
11400
11410
11420
11430
11440
11450
11460
11470
11480
11490
11500
11510
11520
11530
11540
11550
11560
11570
11580
11590
11600
11610
11620
11630
11640
11650
11660
11670
11680
11690
11700
11710
11720
11730
11740
11750
11760
11770
11780
11790
11800
11810
11820
11830
11840
11850
11860
11870
11880
11890
11900
11910
11920
11930
11940
11950
11960
11970
11980
11990
12000
12010
12020
12030
12040
12050
12060
12070
12080
12090
12100
12110
12120
12130
12140
12150
12160
12170
12180
12190
12200
12210
12220
12230
12240
12250
12260
12270
12280
12290
12200
12210
12220
12230
12240
12250
12260
12270
12280
12290
12300
12310
12320
12330
12340
12350
12360
12370
12380
12390
12300
12310
12320
12330
12340
12350
12360
12370
12380
12390
12400
12410
12420
12430
12440
12450
12460
12470
12480
12490
12400
12410
12420
12430
12440
12450
12460
12470
12480
12490
12500
12510
12520
12530
12540
12550
12560
12570
12580
12590
12500
12510
12520
12530
12540
12550
12560
12570
12580
12590
12600
12610
12620
12630
12640
12650
12660
12670
12680
12690
12600
12610
12620
12630
12640
12650
12660
12670
12680
12690
12700
12710
12720
12730
12740
12750
12760
12770
12780
12790
12700
12710
12720
12730
12740
12750
12760
12770
12780
12790
12800
12810
12820
12830
12840
12850
12860
12870
12880
12890
12800
12810
12820
12830
12840
12850
12860
12870
12880
12890
12900
12910
12920
12930
12940
12950
12960
12970
12980
12990
12900
12910
12920
12930
12940
12950
12960
12970
12980
12990
13000
13010
13020
13030
13040
13050
13060
13070
13080
13090
13000
13010
13020
13030
13040
13050
13060
13070
13080
13090
13100
13110
13120
13130
13140
13150
13160
13170
13180
13190
13100
13110
13120
13130
13140
13150
13160
13170
13180
13190
13200
13210
13220
13230
13240
13250
13260
13270
13280
13290
13200
13210
13220
13230
13240
13250
13260
13270
13280
13290
13300
13310
13320
13330
13340
13350
13360
13370
13380
13390
13300
13310
13320
13330
13340
13350
13360
13370
13380
13390
13400
13410
13420
13430
13440
13450
13460
13470
13480
13490
13400
13410
13420
13430
13440
13450
13460
13470
13480
13490
13500
13510
13520
13530
13540
13550
13560
13570
13580
13590
13500
13510
13520
13530
13540
13550
13560
13570
13580
13590
13600
13610
13620
13630
13640
13650
13660
13670
13680
13690
13600
13610
13620
13630
13640
13650
13660
13670
13680
13690
13700
13710
13720
13730
13740
13750
13760
13770
13780
13790
13700
13710
13720
13730
13740
13750
13760
13770
13780
13790
13800
13810
13820
13830
13840
13850
13860
13870
13880
13890
13800
13810
13820
13830
13840
13850
13860
13870
13880
13890
13900
13910
13920
13930
13940
13950
13960
13970
13980
13990
13900
13910
13920
13930
13940
13950
13960
13970
13980
13990
14000
14010
14020
14030
14040
14050
14060
14070
14080
14090
14000
14010
14020
14030
14040
14050
14060
14070
14080
14090
14100
14110
14120
14130
14140
14150
14160
14170
14180
14190
14100
14110
14120
14130
14140
14150
14160
14170
14180
14190
14200
14210
14220
14230
14240
14250

projection of the beam from the point source 88 will be defined by the shape of the aperture in the collimator limited by vertices 92, 94, 96 and 98 and the joining edges, as indicated by lines 130, 132, 134 and 136 in Fig. 7.

5 By adjustment of the source and collimator positions, various projections may be obtained for the image area. An exemplary positioning for these elements is illustrated in Figs. 8 and 9. In the elevational view of Fig. 8, a point source 88 has been rotated off of an orthogonal line between the source and an imaging plane 100. The resulting beam 116, as limited by collimator 14, is projected toward a detector
10 22. In the example illustrated in Fig. 8, the beam 16 has been rotated about an angle 148 with respect to a horizontal line parallel to the image plane. An angle 150 is therefore defined between the inner limit of the beam and the beam center. The point source is located a known distance 152 from the image plane, commonly referred to as the source-to-image distance, or SID. (When the X-Ray beam is
15 angulated, the vertical SID 152 is equal to the actual SID times the cosine of the included angle. When there is no beam angulation, vertical and actual SID are identical.)

Given the geometry set forth in Fig. 8, several angles and distances may be
20 computed. For example, an internal angle 154 is defined between the projected boundary of beam 16 and a plane orthogonal to the beam centerline. Another angle 156 is defined between the image plane 100 and the same plane orthogonal to the beam centerline. Because the distance of the collimator from the point source 88 and the opening in the collimator will generally be known, as will the SID 152, angles 154 and 156 permit computation of the locations at which the beam impinges
25 the image plane 100, as indicated by point 158, point 160, and point 162 in Fig. 8. It should be realized that, while appearing as a point in the projection of Fig. 8, in practice, the impingement locations will typically correspond to lines delimiting the impingement or imaging area. The corresponding distances between these points, as
30 designated by reference numeral 164 and 166 can thereby be computed as well. Finally, as also illustrated in Fig. 8, the computation of the relevant geometries

15-XZ-5566

permits a region 168 wherein the beam impinges the detector surface to be identified for verification purposes as described more fully below.

A plan view of the projection of Fig. 8 is illustrated in Fig. 9. As shown in
5 Fig. 9, the beam 16 propagates, as shaped by the collimator, along a region defined
by lines 130, 132, 134 and 136, to impact the area on the image plane as set forth
above. Where a trapezoidal area such as that illustrated in Fig. 9 is produced by the
beam, it will be noted that the center of the beam impinges the image plane at a
location 160 which is not centered with respect to the other boundaries 158 and 162.
10 In general, however, the projection will be determined by the shape and
configuration of the beam, and the orientation of the source and detector with respect
to the image plane, so as to produce a corresponding image area bounded by vertices
170, 172, 174 and 176 at locations where the corresponding lines 130, 132, 134 and
136 project to the image plane. The image area 178, then, corresponds to the area
15 where the useful image data should be collected, and thereby to the area which can
be cropped for enhanced image processing.

With the foregoing geometry and coordinate systems in mind, analysis of the
relevant data to determine an area to be digitally cropped proceeds generally as
follows. In a present implementation algorithm, position feedback and known
20 geometric information on the relationships between the components, particularly the
source, the detector, and the image plane, are referenced. The spatial points of the
vertices of the collimator field are first computed using coordinate transformation
matrices that are established with angular position feedback of the system. That is,
25 because the angular positions of the source and collimator will be known (e.g.,
sensed), as will their rotational orientations if such freedom of movement is
provided, the locations in space of the vertices of the collimator aperture, denoted by
reference numerals 92, 94, 96 and 98 in the foregoing discussion, will be computed.
Successive rotations or angular displacement of the radiation source about the focal
30 spot or the point source 88 can be managed through appropriate manipulation of the
associated transformation matrices to compute the spatial coordinates of the vertices
in the global (i.e., unrotated) coordinate system.

The present implementation of the transformation matrices will now be discussed with reference to a straightforward application in which the radiation source is rotated about axis 126 (See Fig. 6) by an angle ϕ and the collimator is swiveled about the axis 146 (See Fig.) by an angle φ .

For collimator swivel, the following rigid body transformation matrix is applicable:

$$10 \quad [\varphi] = \begin{bmatrix} \cos(\varphi) & -\sin(\varphi) & 0 \\ \sin(\varphi) & \cos(\varphi) & 0 \\ 0 & 0 & 1 \end{bmatrix},$$

where φ is the angle of rotation of the beam-formatting collimator about a vertical axis Y3 (axis 146 in Fig. 7) through that device (which can be thought of as the central x-ray beam). Similarly, for rigid body rotations of the collimator about the fixed X2 axis (axis 126 in Fig. 6), the following transformation matrix is applicable:

$$15 \quad [\phi] = \begin{bmatrix} \cos(\phi) & 0 & \sin(\phi) \\ 0 & 1 & 0 \\ -\sin(\phi) & 0 & \cos(\phi) \end{bmatrix}.$$

With respect to the coordinate system of the collimator with the origin of the focal point (i.e., the point source 88), each vertex has coordinates that remain constant but that change with respect to the fixed coordinate system. The matrices for these vertices are represented by vector components in the coordinate system 140 illustrated in Fig. 7:

$$25 \quad [P]_i = \begin{bmatrix} P_{1,i} \\ P_{2,i} \\ P_{3,i} \end{bmatrix},$$

where $i = 1, 2, 3, 4$ are the coordinates of the corners 92, 94, 96 and 98 of the collimator aperture, respectively. Similarly, a vector exists for the central ray of the system within the collimator. The actual components of $[P]$ depend upon the basic geometry of the collimator as well as the field size set for the diagnostic examination.

The ensuing computational process involves determining the coordinates of the rotated vectors in the initial unrotated coordinate system. This calculation can be represented by the equation:

10

$$[P] = [\phi][\varphi][P]_i$$

The computed components represent direction values that are unique for each directed line segment (lines 130, 132, 134 and 136 in the foregoing discussion). These direction values are then used to establish the parametric form of the lines in three-dimensional space to determine the intersection of the lines with the image plane of the detector. The computational algorithm, in a present implementation, uses the following series of equations:

$$\frac{X_1 - X_1^0}{P_{1,i}} = \frac{X_2 - X_2^0}{P_{2,i}} = \frac{X_3 - X_3^0}{P_{3,i}}.$$

By selecting the origin of coordinates to coincide with the focal point (i.e., the point source 88 discussed above), the values X_1^0 , X_2^0 , and X_3^0 become zero.

25

It should be noted that this restriction can be removed to assess the affects of non-coincidence between the focal spot and the rotational center of the diagnostic source assembly. Utilizing the source-to-image feedback distance from the system (distance 152 in the projection of Fig. 8), the variable X_3 becomes a known value in the series of parametric equations (equal to the SID), and the intersection coordinates X_1 and X_2 can be computed. Similarly, the predicted intersection point

30

of the center ray of the beam with the detector or image plane can be computer with the associated direction values.

As will appreciated by those skilled in the art, the foregoing approach permits the location of the intersection points of the x-ray beam with the image plane to be computed based upon the known geometry of the system. These points, corresponding to points 170, 172, 174 and 176 in the example illustrated in Fig. 9, provide the parameter vertices for the image to be cropped. Where desired, then, the detector or the detector control circuitry discussed above can command sampling of rows and columns only corresponding to the area bounded by these vertices and their joining lines. Alternatively, additional information can be sampled, but only information relating to this portion of the detector surface need be processed.

As noted above, the present technique also permits comparison of the actual position of the detector with respect to the image area provided by the beam. This may be desired, for example, to establish whether the predicted vertices of the x-ray beam actually impinge the detector. Because the detector and the diagnostic radiation source assemblies are independent, it is possible for them to be spatially misaligned. In the present implementation correlation between the detector position and the beam impingement area is accomplished by comparing the predicted positions of the beam impingement area to the detector imaging surface location to establish whether the image area within the physical confines of the detector.

To implement this feature of the technique, boundaries of the detector are established with respect to the calibrated center of the detector in accordance with the relationships:

$$X_d - W/2 \quad X_{imp} \quad X_d + W/2,$$

and

$$Y_d - W/2 \quad Y_{imp} \quad Y_d + W/2,$$

where X_d and Y_d represent the extremities of the image area, W represents the detector width, and X_{imp} and Y_{imp} represent the coordinates in the image plane where the center of the beam impinges the image plane.

5 With the preceding information the algorithm is capable of computing whether the impingement or image area corresponds to the limits of the detector (i.e., is encompassed by the detector imaging surface). Where such is not the case, an operator may be notified via an alert to correct the detector position or the size and orientation of the projected image area, or the exposure may be inhibited.

10

The foregoing process is set forth diagrammatically in the flow chart of Fig. 10. In the logic of Fig. 10, denoted generally by the reference numeral 180, the SID is first set as indicated at reference numeral 182. At step 184, the source, collimator and detector are positioned, with the source and collimator being angularly positioned and rotated in accordance with the degrees of freedom available in the particular imaging system. At step 186 the position data regarding the angular displacement of the source and collimator, and their rotation, if any, is detected. Such detection may be carried out in any of a range of manners, such as via conventional position sensors. At step 188 the image area is computed as discussed above with reference to the transformation matrices and the SID. With the impingement points in the image plane thus identified, the process may determine whether the image area is within the detector imaging surface bounds as indicated at step 190. As such is not the case, as discussed above, the operator may be notified, or the imaging sequence may be inhibited as indicated at step 192. Once appropriate correction has been made, or if the impingement area is found to be within the bounds of the detector, the desired exposures may be made as indicated at step 194. The image data are then read out and cropped, either during the readout process or subsequently thereto, as indicated at step 196. Finally, the image data may be processed in a conventional manner as indicated at step 198.

20

While the invention may be susceptible to various modifications and alternative forms, specific embodiments have been shown by way of example in

TOP SECRET//SI

the drawings and have been described in detail herein. However, it should be understood that the invention is not intended to be limited to the particular forms disclosed. Rather, the invention is to cover all modifications, equivalents, and alternatives falling within the spirit and scope of the invention as defined by the
5 following appended claims.

U.S. GOVERNMENT PRINTING OFFICE: 1944 10-1200-1